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# A comparison between simulated decoupling regimes for SAR prediction in parallel transmit MRI

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THE CIRCUIT CO-SIMULATION CODE DISCUSSED IN THE PAPER IS FREELY AVIALABLE AT <https://github.com/mriphysics/circuit-cosimulation>. PLEASE CITE THIS WORK IF THE CODE IS USED IN ANY PUBLICATIONS.

## Abstract

**Purpose:** The use of EM modelling is critical for SAR characterisation in parallel transmission MRI. RF arrays that include decoupling networks can be difficult to characterise accurately in simulation. A practical method of simplifying modelling is to exclude the decoupling networks and model each transmit element in isolation. Results from this type of model can be related to a real device by applying ‘active decoupling’ to the real device to suppress residual coupling when in use. Here we compare this approach with a full model that includes decoupling networks.

**Methods:** EM simulations for a variety of adult male voxel models placed within an 8-channel TEM array tuned for 3T operation were run with and without decoupling networks included. The resulting EM fields and SAR estimates were compared using basic normalisation, and simulated active decoupling.

**Results:** Modelling the transmit elements independently leads to variations which have significantly different SAR estimates of ~20% on average compared with the full model if not normalised appropriately. After “active decoupling”, SAR was still generally seen to be overestimated by ~7% with independent channel modelling; despite having similar  $B_1^+$  fields distributions.

**Conclusion:** Modelling transmission elements independently may lead to substantially incorrect SAR estimates if the corresponding MRI system is not run in an analogous manner.

**Key words (abstract):** SAR; EM modelling; parallel transmission MRI; safety

## Introduction

Parallel transmission MRI, in which there are multiple independently driven transmit elements that together generate radiofrequency (RF) electromagnetic (EM) fields, provides new degrees of freedom to both tailor the RF to the subject and to create fields that can be both spatially and temporally varying during a sequence or during individual pulses (1–3). As for any RF transmit system to be used for MRI, the correct characterisation of specific absorption rate (SAR) is paramount to ensuring appropriate safety margins are maintained (4,5). In this regard parallel transmission is both a cause for concern and an opportunity – the user has spatial control over RF electric fields, and this can lead to elevated SAR – but careful optimisation can also lead to configurations in which whole body and in particular maximum local SAR are reduced (1,6). Prior knowledge of the E-field is however required to achieve this, and direct in-vivo measurement is not currently possible using MRI. Some groups have made progress on inferring electric fields and SAR (7,8) from measurements of

the RF magnetic field ( $B_1$ ) but these methods are not yet efficient or accurate enough for practical use. Instead the most common approach is to simulate the RF coil and subject – in the form of a voxelised digital model – using EM field solver software. Effective modelling of a system can enable accurate characterisation of SAR for a given configuration.

Small inaccuracies in the simulation (particularly material properties) mean that if the coil model is constructed using lumped components whose impedance values are taken directly from those in the physical device, the simulated and physical behaviours generally do not match. As a result, lumped element impedances in the model must be altered, typically by adjusting them iteratively. This process is time consuming if full EM field simulations are required at each iteration and is especially troublesome if the simulated coil includes decoupling networks since these contain multiple components to adjust, whose properties affect the responses of multiple transmit elements. A practical solution is to model the individual transmit elements independently without including any decoupling networks (9–11), leading to a system model with “idealised decoupling” in which all elements but one are detuned by removing their lumped capacitors. This is relatively simple to implement since each transmission element can then be tuned and matched independently, however the coupling present in the real system is not represented. In order to use these simulations in practice, the model and real device must be reconciled. This can be achieved by estimating the coupling present and applying it to the idealised model, or by adapting the device itself to approximate the idealised situation as closely as possible. The latter is referred to as “active decoupling” (12–14) and can be achieved by measuring residual coupling, and then driving linear combinations of transmit channels so as to produce a response emulating the behaviour of the coil if no coupling was present between elements. Pickup coils have been used for measurement of residual coupling (13), but the method is not limited to this exclusively.

Iterative tuning of EM models can be made much more efficient by using circuit co-simulation (15). In this approach, the behaviour of the system is first calculated at each lumped element location from multiple full EM simulations. Once characterised in this way, the system behaviour for any arbitrary set of component impedances can be calculated without having to run any further simulations. This approach enables complex models to be tuned using numerical optimisation, with computation time per iteration on the order of milliseconds rather than days. In this way a coil with extensive decoupling networks can still be modelled in full, obviating the need for taking an “idealised” approach. A previous study performed by another group used circuit simulation to investigate the effects of tuning condition, head size and position on SAR of a 9.4T array (16). In this work we consider the

specific example of an 8-channel body coil at 3T, and compare SAR predictions in several different voxel models (9) from “Full Coil Model” simulations generated using circuit co-simulation with “Idealised Coil Model” simulations for the same models generated by excluding decoupling networks and treating each coil element in isolation. To make a direct comparison, the role of active decoupling was also simulated using two different approaches to model how this process may occur in the real world – directly fitting the  $B_1^+$  fields in the isocentre of the models and simulating pickup coils near each transmit element.

## Methods

The system modelled was a 3T Philips Achieva MRI scanner fitted with an 8-channel body transmit coil as described in (12). Simulations were performed using the time domain Finite Integration Technique of CST Microwave Studio (CST AG, Darmstadt, Germany) with a 47 ns Gaussian pulse used for excitation and -50 dB energy decay used for the convergence criterion of the 3D EM simulation (17–19). Simulations were run in a frequency range from 50 to 200 MHz. All other calculations were performed in MATLAB (MathWorks, Natick, MA). All conductive elements were modelled as lossy Copper metal with a conductivity of  $5.8 \times 10^7 \text{ Sm}^{-1}$ .

The simulation was run with *all* lumped elements modelled as  $50 \text{ } \Omega$  S-parameter ports as shown on the left side of Fig. 1. By using circuit co-simulation (15) (as detailed below) the results of this simulation could be used to model the behaviour of the coil for any lumped element impedance values, including removing them by setting a very large impedance. The full coil model has 128 ports of which 120 represent lumped capacitors, with 13 per channel and 16 in the ‘decoupling ring’ structure (see Fig.1). Replacing *all* lumped elements with ports was also found to speed up the simulations in practice by making the coil non resonant, and therefore converge to a steady state for each port excitation far more quickly within the simulation (18).

### Circuit Co-Simulation

Circuit co-simulation has been covered by Kozlov (15) and others (20,21) and is available in commercial software packages. The method can also be implemented directly in a straightforward manner; in this work it was implemented in MATLAB so that it could be combined with MATLAB’s optimisation procedures, a description of the steps utilised is provided here for clarity.

The full 3D EM simulation yields fields and S-parameters (22) for all 128 ports in the coil model: to perform the circuit calculation we must relate these S-parameters – and eventually the measured fields – to those expected for the ports that are actually driven, when a set of defined impedances are placed across the ports corresponding to lumped elements. Starting with the standard definition of a matrix  $\mathbf{S}$  operating on an incident wave  $\mathbf{a}$  (22) producing a reflected wave  $\mathbf{b}$ ,

$$\mathbf{b} = \mathbf{S} \cdot \mathbf{a} \quad (1)$$

we may subdivide this relationship into terms that correspond to the actual physical ports of the system and those that correspond to lumped elements (in our case, all capacitors) (20):

$$\begin{bmatrix} \mathbf{b}_{ports} \\ \mathbf{b}_{lumped} \end{bmatrix} = \begin{bmatrix} \mathbf{S}_{np \times np} & \mathbf{S}_{np \times nl} \\ \mathbf{S}_{nl \times np} & \mathbf{S}_{nl \times nl} \end{bmatrix} \cdot \begin{bmatrix} \mathbf{a}_{ports} \\ \mathbf{a}_{lumped} \end{bmatrix} \quad (2)$$

where there are  $np$  driven ports and  $nl$  lumped element ports (for our model,  $np=8$ ,  $nl=120$ ). The vectors are arbitrarily ordered such that the physical ports occupy indices 1 to  $np$  (compactly referred to via subscript “*ports*”) and the lumped element ports occupy indices  $(np+1)$  to  $(np+nl)$  (referred to via subscript “*lumped*”). A further relation for the lumped element ports is obtained:

$$\mathbf{a}_{lumped} = \mathbf{\Sigma} \cdot \mathbf{b}_{lumped} \quad (3)$$

$\mathbf{\Sigma}$  is another scattering matrix, but the definitions of incident and scattered waves are reversed here because the reflected wave from each port becomes the incident wave on the lumped element with which the port is terminated (21). Unlike  $\mathbf{S}$ ,  $\mathbf{\Sigma}$  is not obtained from calculations made in the original EM simulation, but instead is produced by transformation of the Z-matrix (22) of the lumped elements that are to replace the relevant ports:

$$\mathbf{\Sigma} = (\sqrt{y}\mathbf{Z}\sqrt{y} + \mathbf{I})^{-1}(\sqrt{y}\mathbf{Z}\sqrt{y} - \mathbf{I}) \quad (4)$$

where  $\mathbf{I}$  is a size  $nl \times nl$  identity matrix and  $y$  is the *characteristic* admittance  $1/Z_0$  of each port where  $Z_0$  is  $50 \Omega$  in our case.

By assuming that the lumped elements are completely isolated from one another, entries of  $\mathbf{Z}$  in a system in which all lumped elements are capacitors (as in our case) are defined as:

$$\mathbf{Z}_{n,n} = R - \frac{j}{\omega C_n} \quad (5)$$

where  $C_n$  is the value of the  $n^{\text{th}}$  capacitor with arbitrary numerical index  $n$  and  $R$  is a small (i.e.  $< 1\Omega$ ) series resistance added to reflect both losses within the capacitors themselves and losses in the conductors. This approach has been used by others (23,24) in recognition of the fact that skin effects and other losses are not accurately captured by discretised EM solvers. The value of  $R$  was set empirically by comparing simulation with experimental measurements.

These relations can now be combined to give the  $np \times np$  physical S-matrix  $\mathbf{S}^{phys}$  that would be measured in reality at the driven ports:

$$\mathbf{S}^{phys} = \mathbf{S}_{np \times np} + [\mathbf{S}_{np \times nl}] \cdot \mathbf{\Sigma} \cdot [\mathbf{I} - \mathbf{S}_{nl \times nl} \cdot \mathbf{\Sigma}]^{-1} \cdot [\mathbf{S}_{nl \times np}] \quad (6)$$

The relations above (Eqs.1-6) are written for a single frequency; as we used a broadband simulation, they were applied to all frequencies in the original range. With this framework it was possible to perform a numerical optimisation to find ideal lumped element values to tune, match and decouple the system.

### Tuning and matching

Tuning and matching were performed with the NORMAN male voxel model (25) placed inside the coil in a heart-centred configuration. The physical S-matrix was constructed using the relation detailed in Eq. 6 and appropriate capacitor values were obtained by minimising:

$$\arg \min \left\{ \left\| \mathbf{S}_{i=j}^{phys, \omega_0} \right\| + \lambda (\max \left\{ \left\| \mathbf{S}_{i \neq j}^{phys, \omega_0} \right\| \right\}) \right\} \quad (7)$$

where superscript  $\omega_0$  indicates that the optimisation only considered the desired resonant frequency  $\omega_0$  (128 MHz at 3T). The indices of the  $np \times np$  matrix  $\mathbf{S}^{phys}$ ,  $i$  and  $j$ , relate to the individual transmit channels. Parameter  $\lambda$  was used to adjust the relative weighting of coil matching ( $\mathbf{S}_{i=j}^{phys}$ ) and worst case coupling between transmission elements ( $\max(\mathbf{S}_{i \neq j}^{phys})$ ). The cost function has many local minima; to adequately search the solution space we used the global optimisation algorithm SOMA (26) (Self-Organizing Migrating Algorithm) run with bounds on the capacitor values based on those indicated in (12). Once a suitable local minimum was identified, results were refined using an unconstrained nonlinear optimisation (MATLAB's `fminsearch` function).  $\lambda$  was manually adjusted in both optimisation steps until a suitable solution was found – this was defined as having all elements tuned to 128 MHz, with matching and decoupling of all elements better than -15 dB. Measured S-parameters of the

physical coil can be found in ref. (12). The final model with these optimised capacitor values is known as the Full Coil Model (FCM).

Once suitable capacitor values were determined, the relevant fields were generated by first inferring the power waves present at each lumped element port when each driven port is excited:

$$\mathbf{a}_{lumped} = \mathbf{\Sigma} \cdot [\mathbf{I} - \mathbf{S}_{nl \times nl} \cdot \mathbf{\Sigma}]^{-1} \cdot [\mathbf{S}_{nl \times np}] \cdot \mathbf{a}_{ports} \quad (8)$$

Here  $\mathbf{a}_{ports}$  is a vector of length  $np$  with unit amplitude for the driving port under consideration and zeros elsewhere. Once the full vector  $\mathbf{a}$  was constructed it was then used to calculate the combined E and B fields from the fields generated by every port excitation ( $\mathbf{B}_k$  and  $\mathbf{E}_k$  for port with index  $k$ ):

$$\mathbf{B}_{combined} = \sum_{k=1}^{np+nl} \mathbf{B}_k \cdot \mathbf{a}_k \quad \text{and} \quad \mathbf{E}_{combined} = \sum_{k=1}^{np+nl} \mathbf{E}_k \cdot \mathbf{a}_k. \quad (9)$$

### Idealised Coil Model

Starting from the results of the same EM simulation, the fields produced by a single “ideally decoupled” transmit element were modelled (Fig. 1) by setting all capacitors in other transmit elements to values of  $\sim 10^{-50}$  F (27) (effectively making them open circuit). Each transmit element was then tuned independently using a combined optimisation over several voxel models to find relevant capacitor values – this was done to ensure that the matching for any given model was not overly ideal and hence unrealistic. The results are equivalent to running a full simulation with those capacitors simply deleted from the model as in (9) to produce the “Idealised Coil Model” (ICM) (Fig 1, right hand side).

### Voxel Model Simulations

Simulations using both decoupling regimes were run using an enlarged version of the NORMAN voxel model (28), the virtual population Fats (29) male voxel model and the conventional NORMAN model as shown in Fig. 2a. All models were positioned heart-centred within the coil; an additional simulation was run with the NORMAN model in a head-centred configuration.

The fields were extracted from the simulation software on the calculation mesh (an irregular grid with  $\sim 5\text{mm}$  resolution on average – ranging between 0.5 and 5.2mm with a very fine grid in regions including electrical components and coarser grid in regions with homogenous materials) and the field distributions recovered using Eqs. 8 and 9. All SAR calculations were



performed on the same grid with local 10g averaged Q-matrices (13) calculated for both sets of electric fields – the idealised and full coil model – for each voxel model. The 10g averaging was performing using an in-house algorithm which conforms to the IEEE C95.3 standard (30). The averaged Q-matrices were then condensed down into a much smaller set of Virtual Observation Points (31) (VOPs) with a 1% overestimate bound for SAR comparison. An average compression factor of 2700 was achieved across all the simulations. Using these VOPs, maximum local SAR estimates were calculated for a set of 10,000 random, complex drive settings all with a unit norm.

### Field Normalisation & Active Decoupling

All simulations (both ICM and FCM; all voxel models) were first normalised such that the RF transmit field ( $B_1^+$ ) when each transmit channel is driven independently had mean amplitude 1 $\mu$ T and mean phase of 0° in a central region of interest (5cm<sup>2</sup>) on a slice at isocentre. This is referred to as ‘quadrature’ normalisation since the contributions from each coil were in-phase at the centre of the subject.

As mentioned in the introduction, if an ideally decoupled simulation has been performed then one way of relating this to a real scenario is to employ ‘active decoupling’ (12,13) on the physical system, in which the physical channels are combined so as to further suppress the effects of the channels which are not being driven. In order to allow rigorous comparison of the modelling approaches, the full coil model was subjected to active decoupling; in this way the FCM with active decoupling acts as a surrogate for a physical system with which the ICM can be compared. The efficacy of active decoupling depends on the decoupling matrix used, so in this work two methods were tested. The first was to fit the  $B_1^+$  fields from the FCM in the central transverse slice of each voxel model (excluding regions outside the body) to their ICM counterparts using a least squares fit. This method allows a close as possible match for one set of simulation results to the other but is not feasible for a practical active decoupling implementation.

In practice pickup coils located near each transmit coil can be used to measure residual coupling (12–14). The coil being modelled is made up of transverse electromagnetic (TEM) elements (12) which generate primarily azimuthally orientated magnetic fields between each element and the shield; local pickup coil measurements were simulated by sampling the azimuthal B-field,  $B_\varphi = -B_x \sin(\varphi) + B_y \cos(\varphi)$  in these regions. The simulated pickup coil measurements were collected into matrices of the form:

$$\mathbf{M}_{i,j} = \overline{\mathbf{B}_{\varphi i}(R_j)} \quad (10)$$

containing the mean of  $B_{\varphi i}$  produced by the  $i^{\text{th}}$  coil in a region of interest  $R_j$  near the  $j^{\text{th}}$  coil. Matrices  $\mathbf{M}_{\text{ICM}}$  and  $\mathbf{M}_{\text{FCM}}$  were produced for the idealised and full coil models respectively. The pickup coil actively decoupled FCM was created by applying decoupling matrix  $\text{diag}\{\mathbf{M}_{\text{ICM}}\}\mathbf{M}_{\text{FCM}}^{-1}$  to the FCM. The term  $\text{diag}\{\mathbf{M}_{\text{ICM}}\}$  renormalises the actively decoupled fields to give the same ‘pickup coil readings’ as those from the ICM, making the closest possible match.

In Table 1, we list abbreviations used in this work when referring to the different voxel models and normalisation methods in figures contained in the study.

	Name	Abbreviation
Voxel Model	NORMAN Head-Centred	NHeC
	NORMAN Heart-Centred	NHC
	Enlarged NORMAN Heart-Centred	LNHC
	Fats Heart-Centred	FHC
Simulation Method	Idealised Coil Model	ICM
	Full Coil Model	FCM
	Full Coil Model actively decoupled by fitting	FCM:AD/f
	Full Coil Model actively decoupled by pickup coils	FCM:AD/puc

Table 1: Abbreviations used for different voxel models and different simulation methods

## Results

Decoupling of <-18dB and matching of <-15dB were achieved in the Full Coil Model, as can be seen in Fig. 2b for the heart centred NORMAN model. The two coils with poorer matching are those closest to the arms of the voxel model leading to stronger loading. Better matching could be achieved but this does not reflect the behaviour of the true coil and would lead to an overly idealised situation in which the coil is perfectly tuned for just one scenario but does not behave correctly with different loads. These S-parameters are similar to those measured in the physical coil when loaded (see Fig. 5 in ref. (12)).

Figures 3 and 4 show  $B_1^+$  fields for the heart-centred NORMAN model when just one of the transmit coils (marked in red) is driven in the case of the full and idealised models and with active decoupling to try to match the former to the latter. The full coil model without active decoupling has residual coupling between channels, particularly nearest neighbours (Fig 3a), which is not present at all in the idealised coil model (Fig 3b); as expected the other elements are not even visible in the simulated fields from the ICM. If active decoupling is

applied then agreement improves, particularly when using simulated pickup coils, in which case  $B_1^+$  fields from all but the nearest neighbours of the active coil are suppressed (Fig 3d). Fitting of  $B_1^+$  within the voxel model results in some residual coupling to more distant elements (Fig 3c).

Figure 4 shows the  $B_1^+$  field for the same coil element as shown in Fig. 3, masked within the voxel model only, along with the differences between the ICM and various versions of the FCM for an axial slice through the cardiac region in the NORMAN heart-centred model. Significant differences in magnitude and phase are apparent when comparing the FCM with the ICM directly. Active decoupling leads to closer agreement with the idealised model; as expected, agreement within the subject is best when fitting  $B_1^+$  directly in which case errors are <5%. Active decoupling with pickup coils shows better agreement than quadrature normalisation but does not perform as well as active decoupling by fitting.

The corresponding comparisons for the electric fields show greater differences between the three FCM versions (Figure 5a-c). Although most of the differences in fields from the fitted model are still relatively small, fairly strong variations can be seen in the voxel models' arms where the transmit elements are loaded more strongly, and this leads to greater errors in general. These deviations in the electric field will lead to differences in SAR estimates between the two simulation methods. Furthermore, the electric fields in a different slice show stronger differences (Fig. 5d-f). This is important since SAR must be evaluated throughout the body, not just within the targeted imaging slice(s), in this case one slice at isocentre.

The local SAR estimates from ICM and FCM (calculated from VOP compressed versions of the models) are plotted against each other in Figure 6; in this context the SAR values from the FCM are regarded as being true. The ICM leads to errors of varying degrees depending on the decoupling scheme applied. These differences are summarised by boxplots in Figure 7. It is clear that the plain 'quadrature' normalised FCM leads to the largest discrepancies between the different simulations. Active decoupling by fitting  $B_1^+$  yields the lowest difference although there are still some significant variations, generally with an overestimate for the ICM. Using pickup coils for active decoupling performs in a similar way to active decoupling by fitting. Again there is a tendency for the idealised coil model SAR estimates to be larger.

In order to test whether shim settings resulting in large SAR errors are associated with particular  $B_1^+$  patterns, Figure 8 shows  $B_1^+$  inhomogeneity (coefficient of variation) in the idealised coil model plotted against SAR error (ICM-FCM:AD/f). The two appear to be uncorrelated.

## Discussion & Conclusion

This work compares EM simulations of a parallel transmit MRI body coil using a Full Coil Model consisting of all coil elements plus decoupling circuits with an Idealised Coil Model in which coil elements are treated in isolation. The work focused on a single RF coil model operating at 128MHz but included three separate anatomical voxel models, one of which was simulated in two different positions. The  $B_1^+$  fields predicted by the idealised and full coil models when both are treated with nominal “quadrature” normalisation agree qualitatively but there are large discrepancies in predicted SAR.

In practice an idealised coil model would only be used for comparison with a physical system if some extra step were taken to account for residual coupling – one strategy is to adapt the physical system to emulate the ideally decoupled scenario – this is known as active decoupling (12,13). Active decoupling requires measured estimates of the residual coil coupling; we simulated this measurement by using either (simulated) pickup coils, or by directly fitting the  $B_1^+$  fields. Figure 3 shows that the pickup coil method leads to better suppression of other coil elements than fitting of  $B_1^+$  fields within the body, perhaps because in the latter case the EM fields produced very close to each individual coil element are ignored. As might be expected, directly fitting  $B_1^+$  leads to highly similar  $B_1^+$  in the slice used for fitting (Fig. 4) though errors are not precisely zero. Of course this good agreement is true by construction since we directly fit  $B_1^+$ ; actually the fact that perfect agreement cannot be obtained demonstrates that the ICM and FCM are not simply linear combinations of one another. Active decoupling by fitting also leads to lower median error in simulated SAR predictions (Figs. 6 & 7) than the pickup coils method. This may seem counterintuitive since the fitting is only considering the NMR active  $B_1^+$  component of the magnetic field whereas pickup coils were simulated by considering the B-field produced by currents on the coil elements – as evidenced by the better performance of pickup coils in suppressing fields produced by each element (Fig. 3). Fitting focused on the fields within the body, which are relevant for SAR characterisation – this may explain the better performance. Another possibility is that ensuring similar current distributions on each element for the two decoupling regimes doesn’t necessarily guarantee similar fields *within* the subjects, hence direct fitting of fields yields better results. In practice both methods of active decoupling would be achievable – pickup coils already exist, while fitting could proceed by using in-vivo acquired  $B_1^+$  maps. In reality fitting would be subject to a large degree of uncertainty because of spatial mismatch between the voxel model and acquired  $B_1^+$  data, and so pickup coils will likely remain the preferred method for active decoupling. However it should be

noted that for other types of transmit arrays, pickup coils may not be appropriate – for example, surface coils at 7T and above.

Voxel Model	Quadrature Normalisation					Active Decoupling (fitting)					Active Decoupling (pickup coils)				
	1 <sup>st</sup> Centile	5 <sup>th</sup> Centile	Median	95 <sup>th</sup> Centile	99 <sup>th</sup> Centile	1 <sup>st</sup> Centile	5 <sup>th</sup> Centile	Median	95 <sup>th</sup> Centile	99 <sup>th</sup> Centile	1 <sup>st</sup> Centile	5 <sup>th</sup> Centile	Median	95 <sup>th</sup> Centile	99 <sup>th</sup> Centile
NHeC	-30.5%	-18.4%	21.1%	69.8%	96.2%	-9.7%	-5.5%	3.4%	35.0%	100.3%	-30.0%	-20.6%	5.5%	28.6%	42.6%
NHC	-63.1%	-51.9%	5.1%	173.7%	268.6%	-15.2%	-8.7%	4.5%	36.5%	57.0%	-27.9%	-20.8%	8.2%	41.9%	56.1%
LNHC	-64.5%	-54.0%	-7.6%	119.0%	202.8%	-20.5%	-13.1%	3.5%	36.2%	61.3%	-39.3%	-31.1%	-8.2%	32.2%	50.7%
FHC	-62.7%	-51.7%	4.2%	150.2%	248.0%	-27.4%	-17.1%	3.9%	35.7%	55.2%	-24.7%	-16.2%	12.2%	40.3%	53.6%

Table 2: 1<sup>st</sup>, 5<sup>th</sup>, 95<sup>th</sup> and 99<sup>th</sup> centile of differences along with the mean difference for all voxel models and normalisations

Ultimately it is accuracy of SAR prediction that is the priority for these models. As indicated by the values in Table 2, the most common error if employing an ICM for SAR quantification in conjunction with active decoupling via fitting, is to *overestimate* SAR with the largest overestimate in the heart centred models (99<sup>th</sup> centile of differences) being 61%. The head centred model shows larger percentage differences but these are for relatively very low SAR values so are less of a concern. The data show a general trend towards overestimation that is likely due to greater power dissipation in the subject in the ICM, since there are no other active coil elements for the power to dissipate into elsewhere. From the point of view of safety this makes the models more likely to be conservative, so less of a risk however they would lead to suboptimal operating conditions. Underestimates are more cause for concern. The underestimates were generally of a smaller degree than the overestimates, however the extreme values were still quite large: the maximum underestimate from the ICMs compared with active decoupling by fitting (1<sup>st</sup> centile of differences) was 27%. Median errors in SAR estimation when considering decoupling by pickup coil were slightly larger than for active decoupling via fitting and the peak SAR underestimate in this case was 39%, now for a different voxel model. The implication is that fitting would be a preferred method for active decoupling. However their performance is similar and as was mentioned earlier, though fitting is easy to do in a simulation study like this one, it would be much more error prone if used in reality, and this might introduce an additional source for uncertainty. In general the worst case underestimates and discrepancies between models are of the order of 20-40%, and so margins for error when using such models for safety assessment must be made sufficiently large.

Idealised modelling has been used as a simulation methodology in the past (9,10) primarily due to simplicity of implementation; the models have fewer variables than a full approach and optimisation is more stable due to the lack of coupling. This simulation study has shown that such an approach when combined with active decoupling can lead to SAR predictions which are reasonably accurate, however there are a small number of cases in which larger errors do occur and it is not clear under which conditions we might expect this. In order to explore this Fig. 8 plots the relationship between  $B_1^+$  inhomogeneity and error in SAR estimate, however there does not appear to be a correlation in any model.

The presented results are for one particular case study of an RF coil model – an 8-channel TEM device operating at 128MHz – with findings relatively consistent across a number of anatomical voxel models. It is not immediately clear how these results would generalise to different coils or other frequencies, but a comparison similar to the one made in this work could be applied to different coil geometries and voxel models. Circuit co-simulation can be implemented with commercial software, or using an in-house implementation as outlined here; to assist with this the MATLAB code is freely available at <https://github.com/mriphysics/circuit-cosimulation>. In any case, the simplicity of circuit co-simulation makes it straightforward to model complex devices, perhaps obviating the need for idealised models altogether.

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## Figures

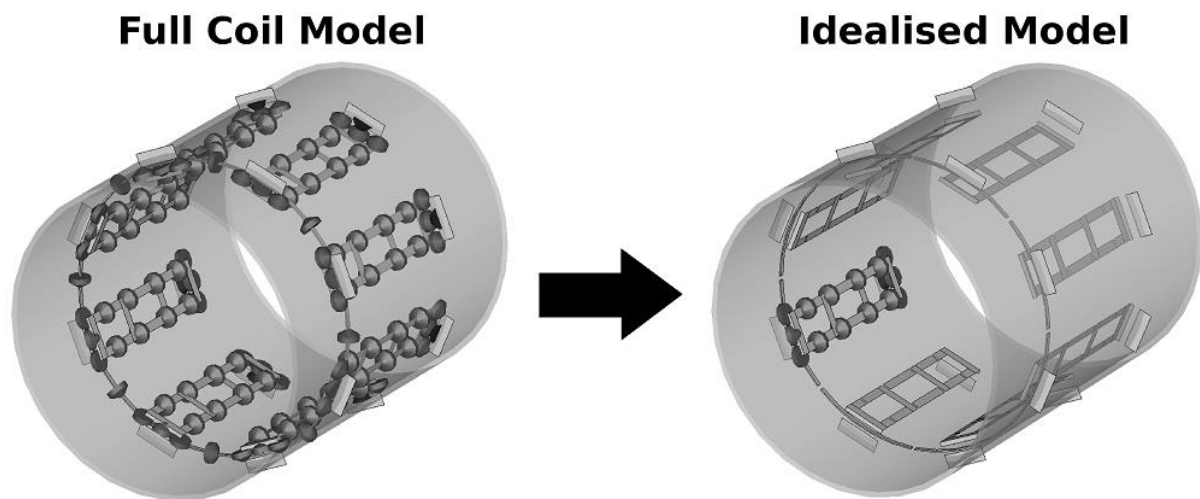


Fig. 1: Schematic of the 8-channel TEM transmit coil. The ring structure at the closer end of the device is used for improving decoupling. The Full Coil Model (left) and Idealised Coil Model (right) differ in that the former includes all lumped capacitors and the latter excludes all those from channels other than the one under consideration. In practice these capacitors can be removed using circuit co-simulation by setting their associated ports in the Full Coil Model to be open circuit (i.e. by assigning a vanishingly small capacitance (27)).

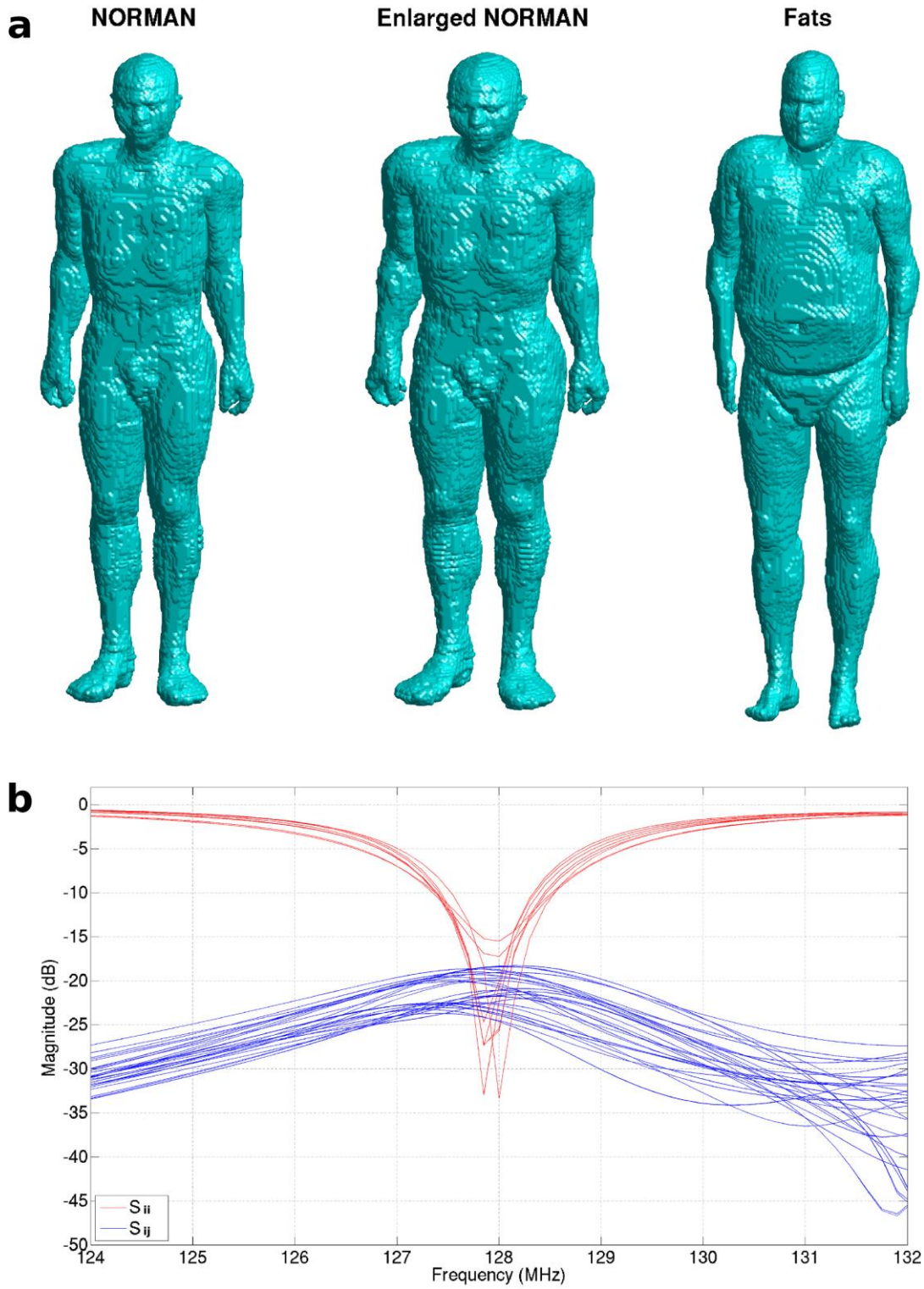


Fig. 2: **(a)** Voxel models used in simulation shown on same scale with BMI of 23.5 for NORMAN, 34.0 for Enlarged NORMAN and 36.1 for Fats. **(b)** S-parameters for the Full Coil Model when loaded with the NORMAN voxel model, heart centred. Plot shows matching and decoupling equivalent to the physical coil ( $S_{ii} < -15$  dB and  $S_{ij} < -18$  dB at 128 MHz)

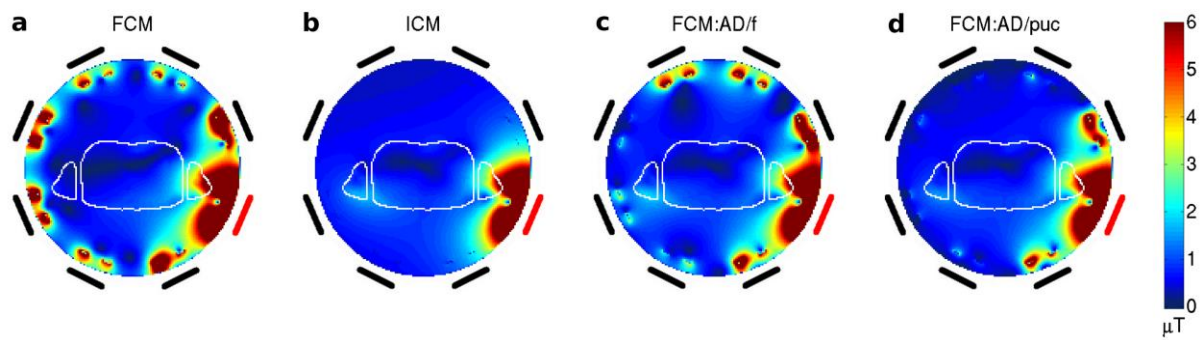


Fig. 3:  $B_1^+$  field magnitudes over the entire coil region for the NORMAN heart centred model in an axial slice through the cardiac region. Relative location of coils is shown outside region with active coil highlighted in red; the outline of the voxel model is plotted in white. The fields from the Full Coil Model are shown alongside those from the Idealised Model and the Full Coil Model with the two different active decoupling methods (by fitting and by simulated coils). Active Decoupling with pickup coils offers the best suppression of alternate transmit channels.

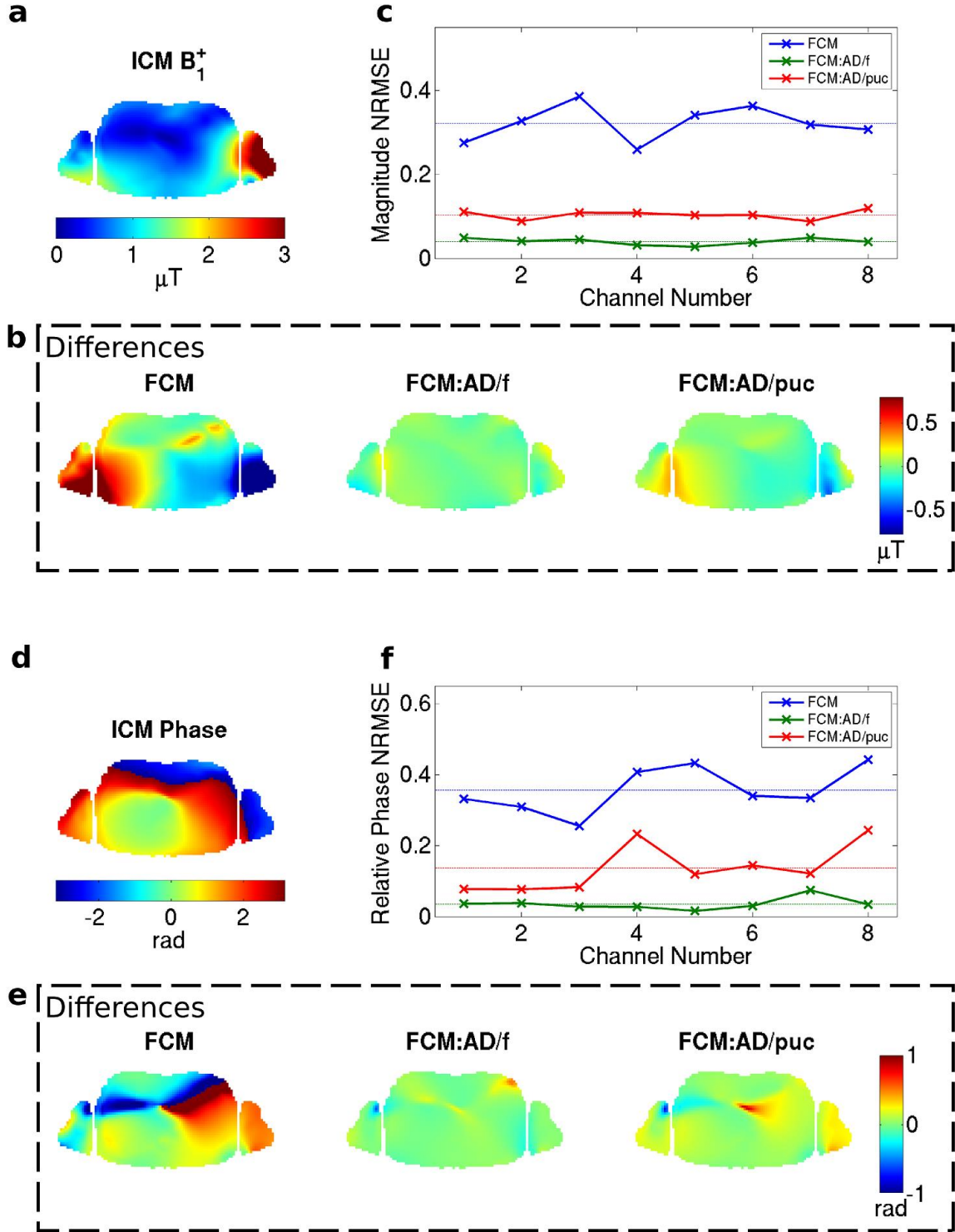


Fig. 4:  $B_1^+$  magnitude in axial slice through the cardiac region of the NORMAN heart centred model (channel 2 shown). (a) Idealised Coil Model (b) Differences from ICM when using FCM with quadrature normalisation, active decoupling by fitting and active decoupling by pickup coils. (c) Normalised RMS difference in  $B_1^+$  between FCM and IM for the eight channels; the dashed lines correspond to averages over all channels. (d-f) As parts (a-c) for phase of  $B_1^+$ .

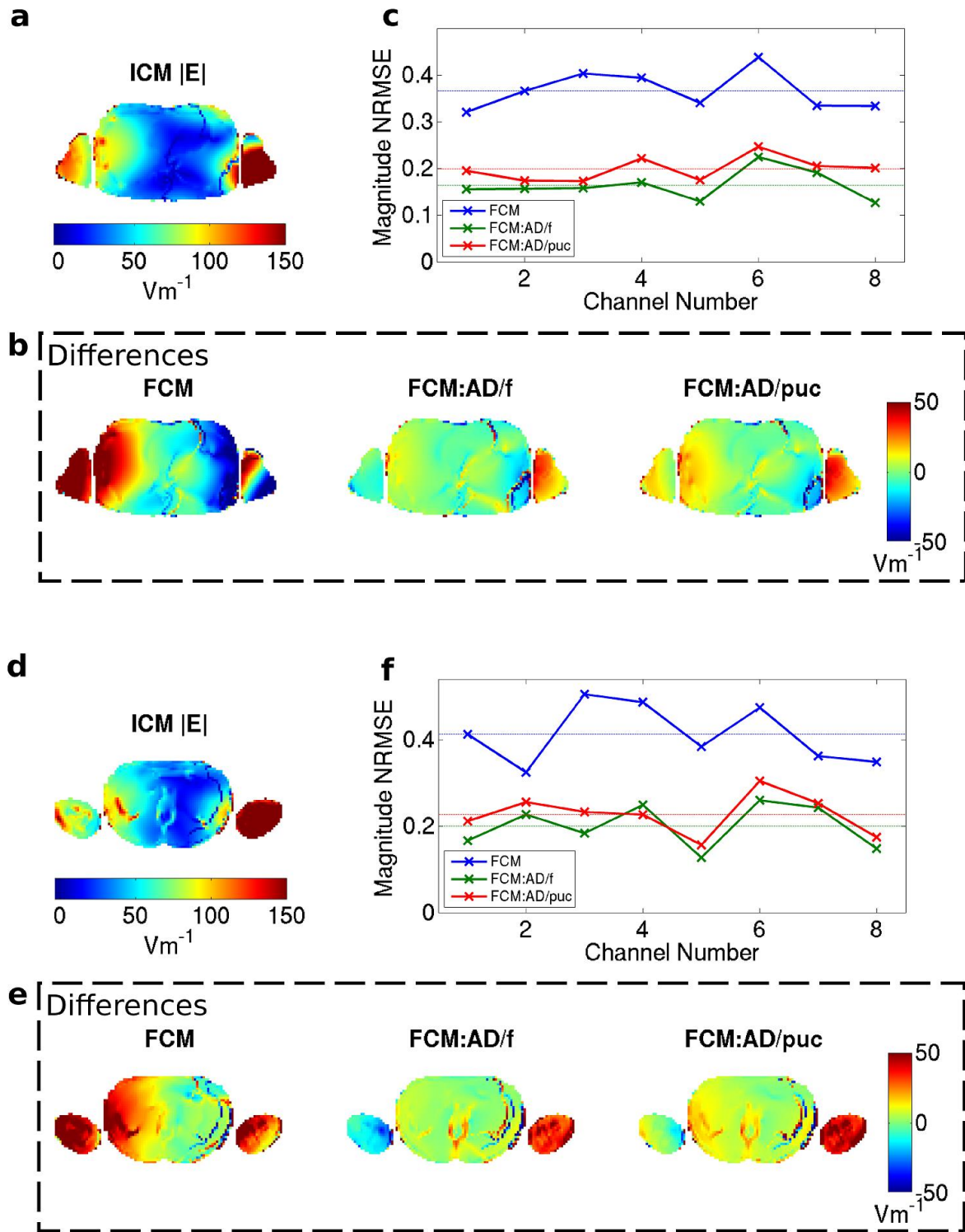


Fig. 5: Magnitude of E-field in the body of the NORMAN heart centred model for channel 2 shown in an axial slice through the cardiac region for (a) the Idealised Coil Model. (b) Differences from ICM when using FCM with quadrature normalisation, active decoupling by fitting and active decoupling by pickup coils. (c) Normalised RMS difference in  $|E|$  between FCM and IM for the eight channels; the dashed lines correspond to averages over all channels. Stronger deviations are seen here compared with the  $B_1^+$  fields for the same slice. (d-f) As parts (a-c) but for a different axial slice in the NORMAN heart centred model – a slice that was not used to calculation the normalisation matrices. Larger differences between the Idealised Coil Model and the Full Coil Model exist in this slice.



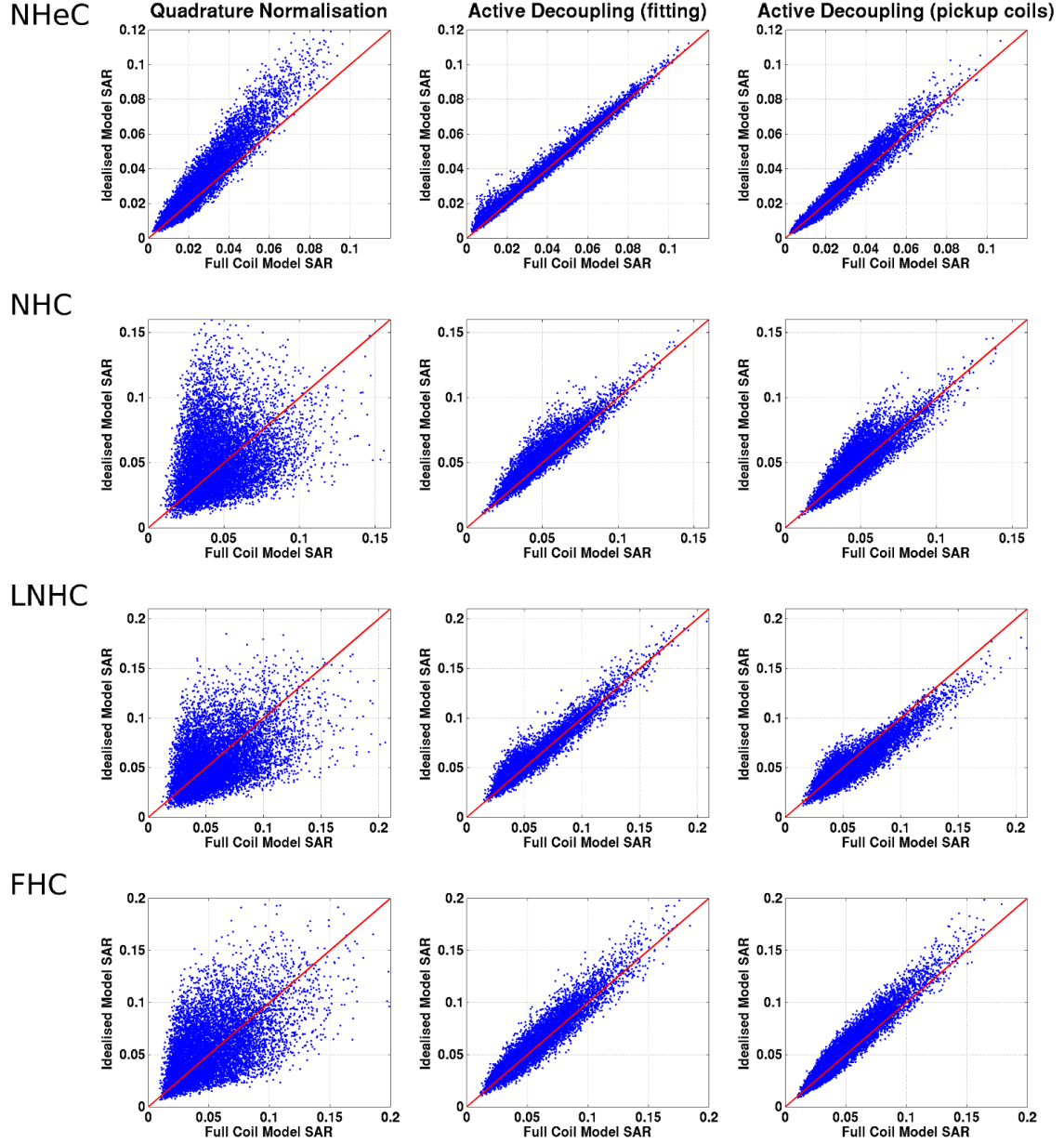


Fig. 6: Maximum local 10g SAR predictions from the Idealised Coil Model plotted against the predictions from the Full Coil Model. Each blue dot corresponds to a single random RF shim; the red line indicates equality. The rows correspond to different voxel models, while the columns correspond to different decoupling settings applied to the FCM (abbreviations listed in Table 1).

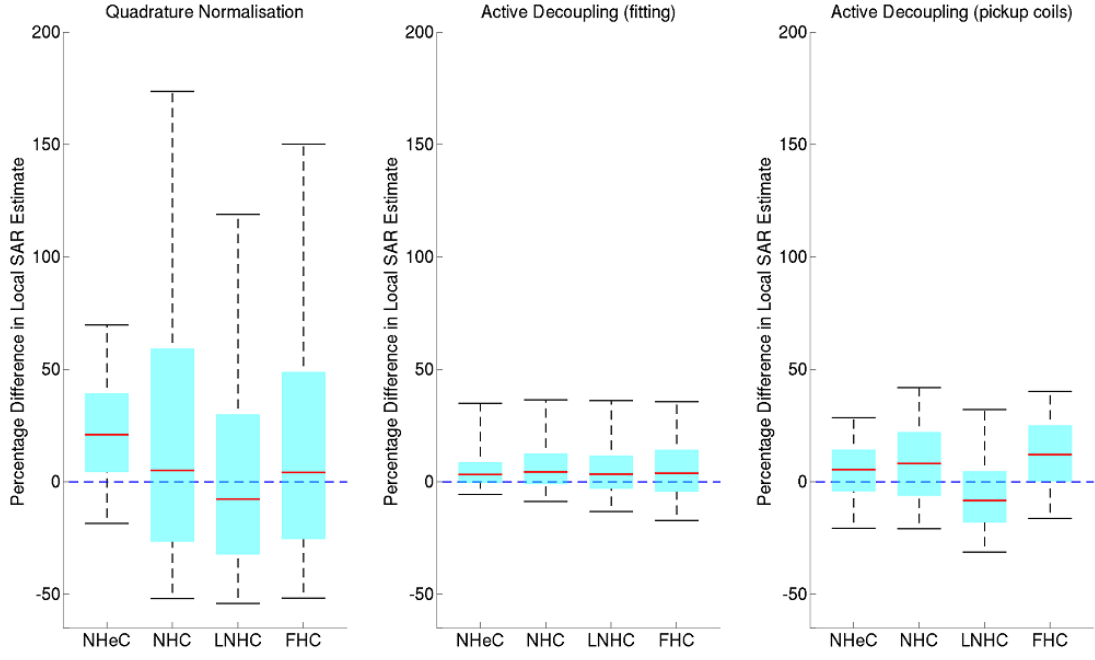


Fig. 7: Boxplots of SAR estimate differences (ICM – FCM) for all three normalisation techniques indicating the 5<sup>th</sup> and 95<sup>th</sup> percentile as the upper and lower bounds and the interquartile range. Positive values indicate overestimation of SAR when using the ICM.

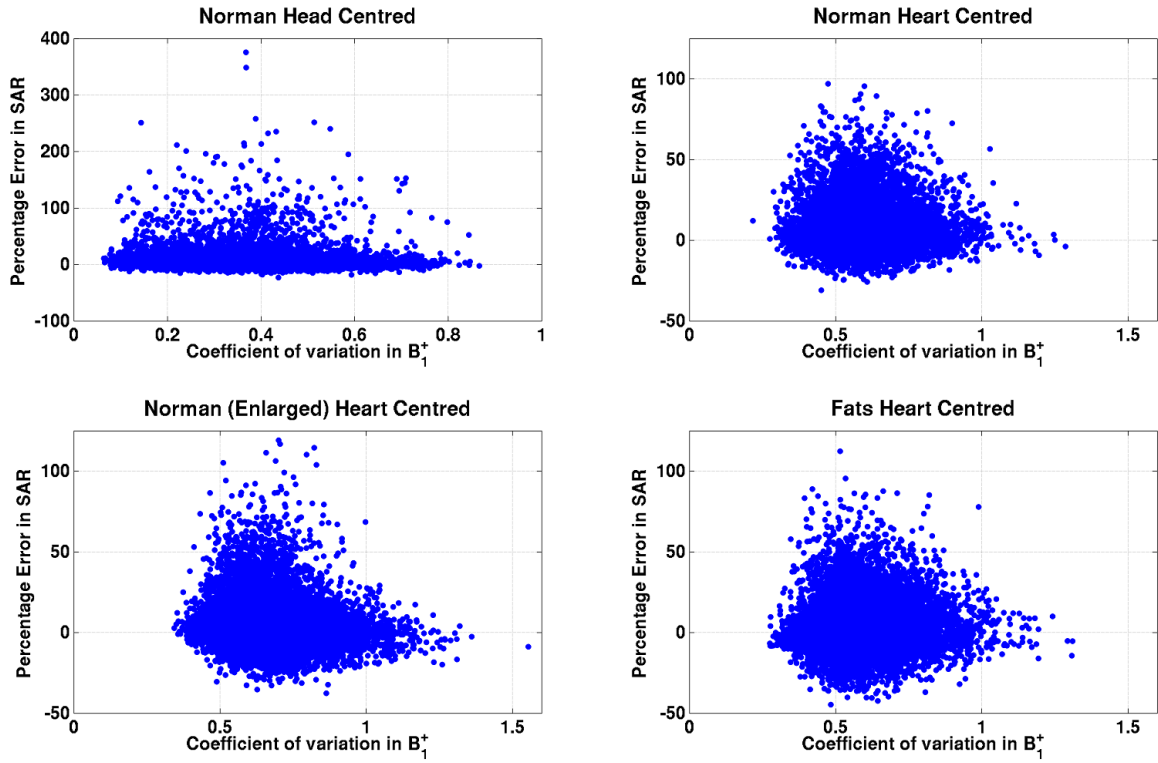


Fig. 8: Coefficient of variation in  $B_1^+$  (high coefficient = poor spatial homogeneity) in the idealised coil model plotted against error in SAR estimate for all voxel models. Each blue dot corresponds to a single random RF shim. There is no clear relation between the two variables.